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The Effect of Load Carriage on Trunk Coordination during Treadmill Walking at Increasing Walking Speed

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Summary

The purpose of this experiment was to determine the effects of walking speed and wearing a backpack on trunk coordination and upper and lower body angular momentum. Twelve subjects (5 male, 7 female, mean age, yr: mean±SD = 26±7.1) walked on a treadmill at increasing speeds from 0.6 $m \cdot s^{-1}$ to 1.6 $m \cdot s^{-1}$ in 0.2 $m \cdot s^{-1}$ increments. Subjects walked with a backpack (BP) containing 40% of their body mass and with no backpack (NBP). Peak pelvic and thoracic angular velocities were measured, and peak upper body and lower body angular momentum and the relative phase between the pelvis and thorax were calculated. A Repeated Measures ANOVA with two within-subject factors (load and speed) was used to compare the dependant variables. A significant main effect of BP condition was found in pelvic ($p < 0.0001$) and thoracic ($p < 0.0001$) angular velocities, upper ($p < 0.0003$) and lower ($p < 0.0001$) body angular momentum, and relative phase ($p < 0.0014$). In addition, a significant main effect of walking speed was found in thoracic angular velocity ($p < 0.0001$), pelvic angular velocity ($p < 0.0001$), upper body angular momentum ($p < 0.0001$), lower body angular momentum ($p < 0.0001$), and relative phase ($p < 0.0001$). A significant interaction effect between speed and load was determined for thoracic angular velocity ($p < 0.0001$), upper body angular momentum ($p < 0.0006$), and relative phase ($p < 0.0001$). There were higher pelvic and thoracic angular velocities, and higher upper and lower body angular momentum in the NBP condition compared to the BP condition. In the NBP condition, relative phase between pelvic and thoracic rotation increased from 54° at .6 $m \cdot s^{-1}$ to 122° at 1.6 $m \cdot s^{-1}$. In contrast, the increase in relative phase in the BP condition was less, from 48° at .6 $m \cdot s^{-1}$ to 78° at 1.6 $m \cdot s^{-1}$. With the addition of the BP, the decrease in thoracic angular velocity observed was disproportionate to the increase in the moment of inertia of the upper body caused by the addition of the pack, resulting in lower upper body angular momentum compared to the NBP condition. The lower upper body angular momentum in the BP condition may have occurred as a means to reduce the muscular force required to rotate the thorax, and hence lowered the metabolic cost of the increased load.

Introduction

Load carriage research performed at the United States Army Research Institute of Environmental Medicine (USARIEM) is designed to investigate how different physical properties of a backpack affect the individual carrying the load. Generally this approach involves isolating and manipulating a specific property of the backpack such as mass, volume, moment of inertia, or position of the center of mass (COM). The next step is to determine how manipulating this property affects variables such as metabolic cost, joint reaction force, and performance measures. For instance, Obusek (4) determined that when the COM of the backpack was high and close to the body, there was a reduced metabolic cost and lower limb joint reaction force. The purpose of the current experiment was to investigate the effects of walking speed and wearing a backpack on transverse plane thoracic and pelvic angular velocity, the angular momentum of the upper and lower body, and trunk coordination across different walking speeds. This study provides the basis for future work to determine the effect of load distribution on the individual carrying the load.

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During unloaded walking, increasing walking speed has been shown to affect parameters of gait such as hip excursion, stride length and stride frequency. Previous research has demonstrated that increasing walking speed is associated with increases in transverse plane pelvic rotation, transverse plane thoracic rotation, and trunk rotation (8, 9). In addition, Wagenaar and Beek (8) showed that at walking speeds less than approximately 0.8 $m \cdot s^{-1}$, the pelvis and thorax rotated in the same direction in the transverse plane (in-phase), while at higher walking speeds (> 1.0 $m \cdot s^{-1}$) the pelvis and thorax rotated in opposite directions (out-of-phase). Literature suggests (5, 9) the out-of-phase pattern between pelvic and thoracic rotation at walking speeds greater than 0.8 $m \cdot s^{-1}$ reduces the net angular momentum of the body.
Researchers have shown that stride frequency and transverse plane pelvic rotation increased with increasing walking speed (9). Taken together these suggest higher levels of pelvic angular velocity would result from higher walking speeds. In addition, it has been shown that due to a larger hip excursion and leg swing, the moment of inertia of the lower body increases at higher walking speeds (6). Because angular momentum is the product of angular velocity and moment of inertia, it was expected that lower body angular momentum would increase with increasing walking speed. Our first hypothesis was that increasing walking speed would cause an increase in pelvic angular velocity and lower body angular momentum.
Adding a backpack increases the moment of inertia of the thorax. Consequently, when carrying a backpack, less thoracic angular velocity will result in comparable levels of upper body angular momentum as when not carrying a backpack. If the thorax rotates in order to counter balance the angular momentum of the lower body, a decrease in thoracic angular velocity is expected in the backpack condition. Therefore, it was hypothesized that during load carriage there would be a reduction in thoracic angular velocity in order to maintain comparable levels of upper body angular momentum, as when not wearing a backpack.
By adding a backpack that makes contact only with the thorax, we did not change the moment of inertia of the lower body. Because an increase in pelvic angular velocity was expected to result from increasing walking speed, an increase in lower body angular momentum was also expected. Similar increases in pelvic angular momentum were expected when carrying a backpack as when not carrying a backpack. Previous research has suggested as lower body angular momentum increases, there is a transition from an in-phase to an out-of-phase pattern between pelvic and thoracic rotation. In the present study it was hypothesized that increasing walking speed would cause a transition from an in-phase to an out-of-phase pattern of pelvic and thoracic rotation regardless of whether or not the subject was carrying a backpack.

Methodology

Volunteers

Fourteen healthy subjects participated in the study. Two subjects were excluded from the final analysis because of technical problems during data collection. The remaining five male and seven female volunteers were used in the final analysis. Subjects were (age, yr: mean±SD = 26±7.1) from the Boston University community, participated in strenuous physical exercise at least three times per week and had no orthopedic disorders or complicating medical histories. Prior to participation, subjects gave informed consent in accordance with the policies of the Boston University Institutional Review Board.

Experimental Procedures

Anthropometric Measures. Data were collected in the Barreca Musculoskeletal Laboratory at Boston University. Prior to data collection, anthropometric measures were taken of total leg, shank, thigh and arm length, as well as hip and shoulder width. Body mass and height were measured using a balance scale.

Experimental Protocol. Subjects walked with and without the backpack at six different speeds. The sequence of backpack or no backpack condition was balanced across subjects. Walking speed was increased from 0.6 $m \cdot s^{-1}$ to 1.6 $m \cdot s^{-1}$ in 0.2 $m \cdot s^{-1}$ increments, and then decreased in the same manner. There were a total of 6 speed conditions for each of the 2 backpack conditions. Subjects walked at each speed for approximately 3 minutes. During the last 30 seconds, kinematic and kinetic data were collected.

Instrumentation

Kinematic and Kinetic Data Collection Systems. Three-dimensional kinematic data were collected at 100 Hz through an Optotrak 3020 System (Northern Digital Inc., 403 Albert Street, Waterloo, Ontario, Canada, N2L 3V2). Cameras were placed on each side of a treadmill, approximately 3 meters from the treadmill. Subjects walked on an instrumented Kistler / Trotter treadmill (Gaitway model; Kistler Instrument Corporation, 75 John Glenn Drive, Amherst 14228-2171) capable of measuring vertical ground reaction force (VGRF). On average, the belt speed varied .01 $m \cdot s^{-1}$ (< 2%) within a speed condition regardless of load. The Optotrak system unit provides an external trigger that was used to trigger the start of the force plate data collection, thereby synchronizing the kinematic and kinetic data.

Optotrak required the use of infrared light emitting diodes (IREDS), which were placed bilaterally on the subject's zygomatic processes, acromion processes, mid thighs, lateral femoral condyles, lateral malleoli, and ulnar styloid processes. Transverse pelvic and thoracic rotations were recorded, using two custom made T-squares (9). IREDS were placed on each end of the T-squares (Figure 1). IREDS were also placed bilaterally on the center of mass of the backpack in the sagittal plane, and on the ends of the bar holding the weight.

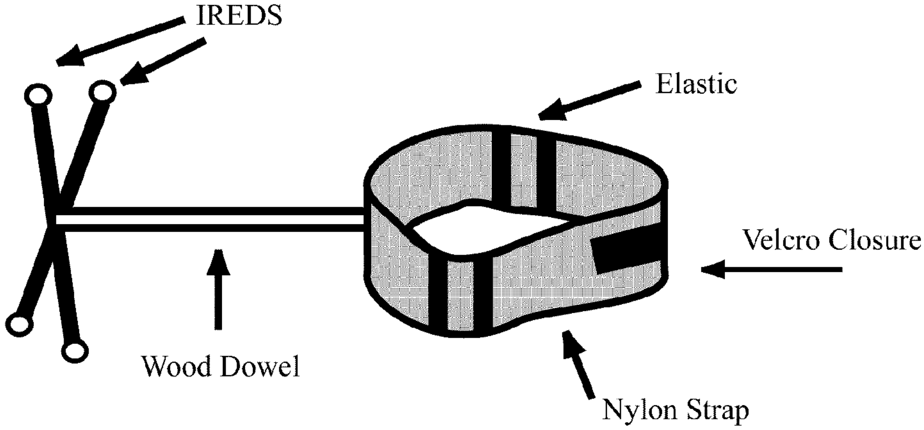


Figure 1. T-Squares

Backpack. The backpack frame was constructed of rigid plastic and designed to make contact only with the thorax (Figure 2). An aluminum rod was attached to the frame to hold a disk shaped weight at shoulder height as close to the subject as possible. Two adjustable shoulder straps and a mid thoracic strap minimized pack movement in relation to the thorax. The total weight of the backpack was adjusted to 40% of the subject's body weight. The weight was chosen to fall in the range of weights normally tested in backpack experiments (2, 4, 7).

DATA PROCESSING

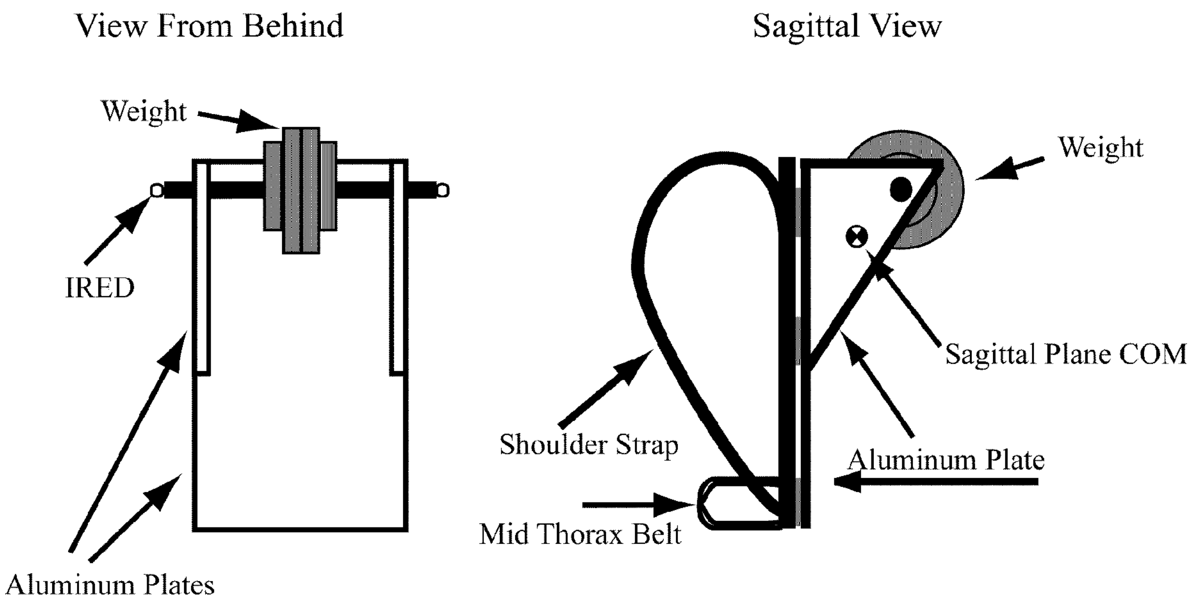


Figure 2. Backpack

Heel strike was determined as the first frame of the time series that the VGRF was greater than 7% of the peak VGRF. Toe off was determined as the last frame of the time series that the VGRF was greater than 7% of the peak VGRF (1). Heel strike and toe off data were used to calculate stride length and frequency.
The raw kinematic data were converted into 3D data by means of the Optotrak system software. Missing data were interpolated using a cubic spline. If there were more than 15 consecutive frames of missing data within any particular stride, that stride was discarded because the interpolation was not reliable. The interpolation procedure was validated against known values before its use and demonstrated a maximum error of 1.0 mm. Aberrant force-plate data would occur if both feet were on the same force-plate at the same time. There were a total of 4753 strides of data collected, of which 3.9% were eliminated due to missing kinematic or aberrant force-plate data. After interpolation, the data were filtered at 5 Hz (low pass second order Butterworth). Pelvic and thoracic rotations in the transverse plane were measured from the interpolated and filtered time series. Angular velocity was calculated from the displacement data. The transverse plane axis of rotation of the pelvis was assumed to be located at the spine, midway between the greater trochanters of the left and right hips. The transverse plane axis of rotation of the thorax was also assumed to be located in the spine, midway between the shoulders.

The moment of inertia of the upper and lower body were calculated using equation 1:

$$MoI = \sum_{i=1}^{n} m_i r_i^2 \qquad (1)$$

m is the mass of the segment. r represents the distance between the center of mass of that segment and the axis of rotation and n represents the number of segments. Segment mass and the position of each segment's center of mass were based on anthropometrics and calculated from estimates given by Dempster (3). In the case of the lower body, there were 5 segments (n =5): 2 shanks, 2 thighs and the pelvis. The mass of the feet was included in the mass of the shank segments. In the case of the upper body, there were 3 segments (n = 3): 2 arms and the thorax. For the purpose of the moment of inertia calculation, the upper and lower arm was considered to be one segment. The upper body also included the mass of the head. The contribution of the backpack to the moment of inertia of the upper body was calculated using the parallel axis theorem. Reported values are peak angular velocity and angular momentum.

Relative phase was used as a measure of coordination between pelvic and thoracic rotation and was calculated using the method described in van Emmerik and Wagenaar (8). An in-phase pattern indicates the pelvis and thorax are rotating in the same direction, and an out-of-phase pattern indicates the pelvis and thorax are rotating in opposite directions. Relative phase was calculated in the interval between 0° – 180°. Although relative phase is a continuous variable, generally an in-phase pattern is characterized by relative phase values between 0° and 90°, while an out-of-phase pattern is characterized by relative phase values between 90° and 180°. Repeated measures Analysis of Variance (ANOVA) with two within-subject effects was used to test for the main effects of speed (six levels) and backpack (two levels) conditions on the dependent variables. If significant interaction effects of backpack and speed were found, data were subset by walking speed and a repeated measure ANOVA was used to test for main effects of backpack at each walking speed.

Results

Effect of Carrying a Backpack

There was a significant main effect of backpack condition on thoracic angular velocity ($p < 0.0001$), pelvic angular velocity ($p < 0.0001$), upper body angular momentum ($p < 0.0003$), lower body angular momentum ($p < 0.0001$), and relative phase ($p < 0.0014$). Averaged across walking speeds, walking with a backpack resulted in a 40% lower pelvic angular velocity and a 54% lower thoracic angular velocity.

Figure 3

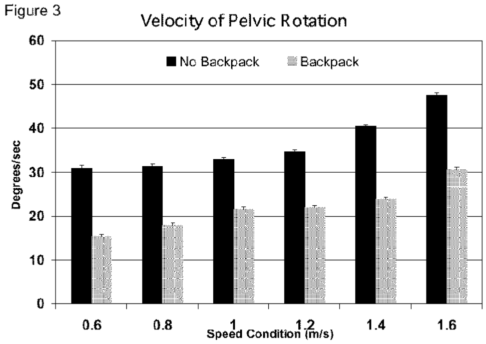


Figure 4

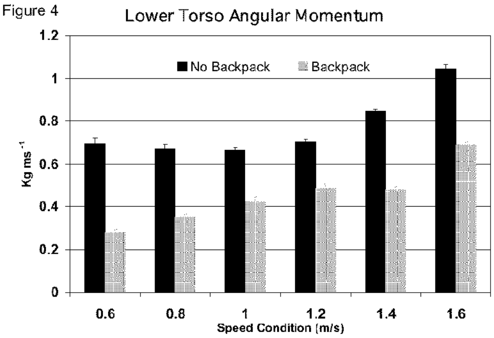


Figure 5

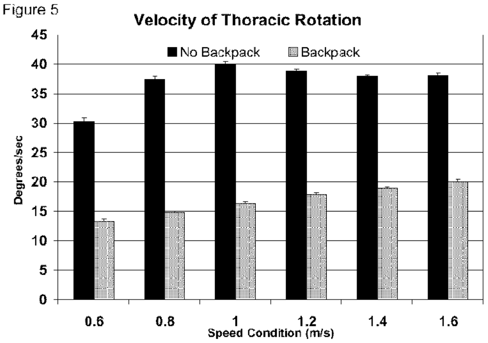


Figure 6

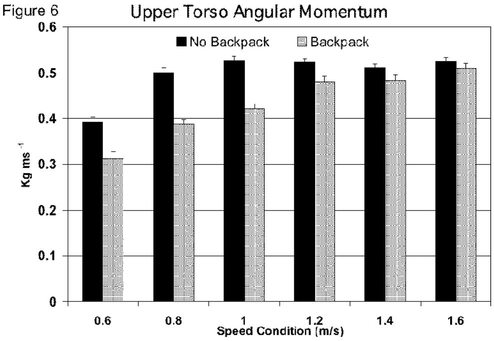
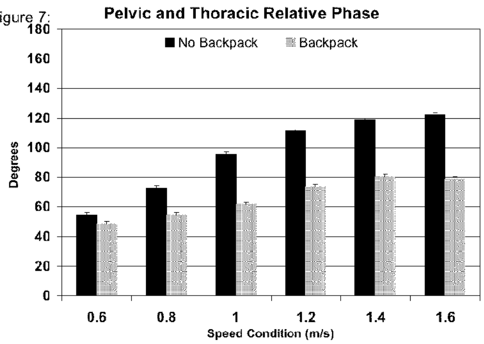


Figure 7:



Similarly, in the backpack condition there was a 38% lower upper body angular momentum and a 13% lower, lower body angular momentum. Adding a backpack resulted in a more in-phase pattern of relative phase (67.87°) compared to the no backpack condition (99.81°). Figures 3, 4, 5, 6, and 7 show the means and standard errors for thoracic and pelvic angular velocity, upper and lower body angular momentum, and relative phase across backpack conditions and walking speeds.

Effect of Walking Speed

A significant main effect of walking speed was found on thoracic angular velocity ($p < 0.0001$), pelvic angular velocity ($p < 0.0001$), upper body angular momentum ($p < 0.0001$), lower body angular momentum ($p < 0.0001$), and relative phase ($p < 0.0001$). Increasing walking speed from .6 $m \cdot s^{-1}$ to 1.6 $m \cdot s^{-1}$ resulted in a 39% increase in pelvic angular velocity, and a 23% increase in thoracic angular velocity. Similarly, lower body angular momentum increased 40% and upper body angular momentum 27% with increasing walking speed. In addition, at .6 $m \cdot s^{-1}$ the relative phase between pelvic and thoracic rotation was 51.78°, while at 1.6 $m \cdot s^{-1}$ the relative phase was 103.35°.

Interaction Between Walking Speed and Backpack Condition

A significant interaction effect between walking speed and backpack condition was determined for thoracic angular velocity ($p < 0.0001$), upper body angular momentum ($p < 0.0006$), and relative phase ($p < 0.0001$). No significant interaction between walking speed and backpack condition was found for pelvic angular velocity ($p < 0.1254$) or lower body angular momentum ($p < 0.0524$). Increasing walking speed in the no backpack condition caused significantly larger increases in thoracic angular velocity and upper body angular momentum compared to increasing walking speed in the backpack condition. In addition, increasing walking speed in the no backpack condition resulted in an increase in relative phase from 52° at .6 $m \cdot s^{-1}$ to 122° at 1.6 $m \cdot s^{-1}$, while in the backpack condition the increase was from 48° to 78°. When data were subset by walking speed, we did not detect significant differences in relative phase at .6 $m \cdot s^{-1}$ between backpack conditions ($p < 0.5076$); however, significant differences in relative phase were detected at 1.6 $m \cdot s^{-1}$ ($p < 0.0001$).

Discussion

In the present study we hypothesized that 1) increasing walking speed would result in an increase in pelvic angular velocity and increased lower body angular momentum; 2) wearing a backpack would result in lower thoracic angular velocity; and 3) increasing walking speed would cause a transition from an in-phase to an out-of-phase relationship between pelvic and thoracic rotation, regardless of backpack condition. Our results indicate increasing walking speed was associated with higher levels of pelvic angular velocity and lower body angular momentum, supporting our first hypothesis. In addition, carrying a backpack was associated with lower levels of thoracic angular velocity and a persistent in-phase pattern of pelvic and thoracic rotation compared to walking in the no backpack condition. Contrary to our predictions, no out-of-phase relationship between pelvic and thoracic rotation was observed in the backpack condition, and there was less upper body angular momentum compared to the no backpack condition.

Consistent with the findings of Wagenaar and Beek (9), increasing walking speed in the no backpack condition was associated with a transition from an in-phase to an out-of-phase relationship between pelvic and thoracic rotation. The transition occurred at walking speeds of approximately .8 $m \cdot s^{-1}$ to 1.0 $m \cdot s^{-1}$. In addition, increasing walking speed in the no backpack condition was associated with an increase in lower body angular momentum. Wagenaar and Beek have suggested that the pelvis and thorax rotate out-of-phase at higher walking speeds as a mechanism for reducing the net angular momentum of the body. When the pelvis and thorax rotate out-of-phase, high levels of lower body angular momentum are counterbalanced by high levels of upper body angular momentum in the opposite direction. However, the lower body angular momentum in the backpack condition at the highest walking speed was comparable to the lower body angular momentum in the no backpack condition at walking speeds, below which there was the transition to an out-of-phase pattern. This suggests the in-phase pattern of pelvic and thoracic rotation may be attributable

to the lower level of lower body angular momentum in the backpack condition compared to the no backpack condition.

The decrease in thoracic angular velocity in the backpack condition was disproportionate to the increase in the upper body moment of inertia from the addition of the backpack mass, resulting in a decrease in upper body angular momentum. The larger moment of inertia of the upper body when wearing a backpack likely requires a greater degree of muscle force in order to rotate the upper body and to generate the angular momentum necessary to counter the opposing pelvic angular momentum. Consequently, maintaining an in-phase pattern between pelvic and thoracic rotation while walking with a backpack may result in a lower metabolic cost.

This study provides insight into the kinematic adaptations associated with carrying a backpack and their kinetic consequences, and offers hypotheses about the causes of these changes. The information can be used by designers to develop packs that reduce metabolic cost and improve soldier performance during load carriage. However, further investigation is needed in order to determine if metabolic cost is affected by changes in trunk coordination resulting from adding a backpack.

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